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Research Article

Sudden stop detection and automatic seating support with neural stimulation during manual wheelchair propulsion

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Objective: Wheelchair safety is of great importance since falls from wheelchairs are prevalent and often have devastating consequences. We developed an automatic system to detect destabilizing events during wheelchair propulsion under real-world conditions and trigger neural stimulation to stiffen the trunk to maintain seated postures of users with paralysis.

Design: Cross-over intervention

Setting: Laboratory and community settings

Participants: Three able-bodied subjects and three individuals with SCI with previously implanted neurostimulation systems

Interventions: An algorithm to detect wheelchair sudden stops was developed. This was used to randomly trigger trunk extensor stimulation during sudden stops events

Outcome Measures: Algorithm success and false positive rates were determined. SCI users rated each condition on a seven-point Usability Rating Scale to indicate safety.

Results: The system detected sudden stops with a success rate of over 93% in community settings. When used to trigger trunk neurostimulation to ensure stability, the implant recipients consistently reported feeling safer ($P<.05$ for 2/3 subjects) with the system while encountering sudden stops as indicated by a 1–3 point change in safety rating.

Conclusion: These preliminary results suggest that this system could monitor wheelchair activity and only apply stabilizing neurostimulation when appropriate to maintain posture. Larger scale, unsupervised and longer-term trials at home and in the community are indicated. This system could be generalized and applied to individuals without an implanted stimulation by utilizing surface stimulation, or by actuating a mechanical restraint when necessary, thus allowing unrestricted trunk movements and only restraining the user when necessary to ensure safety.

Trial Registration: NCT01474148

Keywords: Spinal cord injury, Electrical stimulation, Wheelchairs, Mobility, Safety

Introduction

In the United States, approximately 3.6 million individuals rely on wheelchairs as their primary form of mobility.[1] Wheelchair safety is of great importance since falls from wheelchairs are very prevalent and often have devastating consequences. Over 100,000 wheelchair related injuries occur annually and the vast majority of them result from falls and tips.[2] Up to 73% of wheelchair users experience falls, with 13–38% of wheelchair users being injured, and 5% of users being seriously injured.[3–5] Fear of falling is reported by 73% of wheelchair users[6] and this fear can lead to a decreased quality of life.[7] These falls and tips can result in lacerations, fractures, head injuries, and even death.[2] A large portion of these wheelchair tips are

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due to traveling over uneven terrain (such as a bumpy sidewalk where the caster can fall into a pothole) or colliding into an object (e.g. curb).[8] The most frequent attribute for wheelchair falls has been attributed to terrain condition.[6] Modifications can be made to the wheelchair to allow them to be safer in these environments (larger caster wheels, adjusting the seating position), however this can affect maneuverability and ergonomics.[3] Seatbelts or straps can also secure the rider when sudden stops or impacts occur, however many users reject such restraints due to the difficulties of donning and doffing, the restrictions they place on stationary trunk movements, and their potential contribution to more severe complications including pressure ulcers, skin breakdowns, and asphyxiation.[9,10] Transitioning to a power wheelchair may help since they offer a wider availability of seating configurations, however doing so can limit accessibility, incur the social stigma of using a device that implies a more severe disability[11] and still may be susceptible to injurious falls in some situations.[12]

A large percentage of manual wheelchair users consists of individuals with spinal cord injury (SCI). In the United States, the SCI population is estimated to be approximately 288,000, with 17,700 new cases each year.[13] After an SCI, many of these individuals lose the ability to activate the postural muscles of the hips and trunk to position and stabilize themselves while in a seated position. The paralysis of the core trunk and hip muscles can lead to a decreased work volume,[14] inefficient manual wheelchair propulsion[15,16] and difficulties during many activities of daily life which can lead to chronic pain.[17] This population is also susceptible to falls and injuries during wheelchair propulsion due to the inherent instability and lack of ability to recover from external perturbations.

Smart wheelchairs[18] utilizing methods involving inertial measurement units (IMUs),[19–21] stereoscopic cameras,[22,23] and LIDAR (Light Detection And Ranging)[24,25] to help detect terrain conditions and drive autonomously are rapidly becoming popular. IMUs are also being used for activity monitoring and to measure active manual wheelchair propulsion.[26–28] Neural stimulation has been shown to be an effective means of trunk control for seated stability[29–31] and wheelchair propulsion[32,33] for individuals with SCI. Systems employing an IMU to detect destabilizing wheelchair events and activate appropriate neural stimulation to maintain or restore seated posture have been demonstrated in laboratory settings.[34,35] In Crawford *et al.*,[34] a post-processing analysis of various events and terrains in a laboratory setting that classified destabilizing events from trunk and wheelchair mounted accelerometers showed the potential to use these signals to control for neural stimulation to stabilize the posture. In Armstong *et al.*,[35] sudden stops and turns of consistent velocity and energy just below those predicted to cause a fall were applied by guiding manual wheelchair users down a fixed incline with a central rail and roller bearings. Collisions and 90° turns were successfully detected by a wheelchair mounted IMU and used to trigger neural stimulation to the gluteal, erector spinae and quadratus lumborum muscles, which significantly reduced forward or lateral trunk lean and time to recover to an erect sitting posture after the events. However, both of these investigations were conducted in carefully controlled conditions rather than real-world situations. The work presented here expands upon these proof-of-concept studies to detect unexpected and potentially destabilizing wheelchair events in community settings and activate the appropriate stimulation to stabilize the user and provide for safer propulsion in real world environments without restrictive physical restraints or other wheelchair modifications.

Methods

Study design

The study consisted of three major phases – algorithm development, able-bodied (AB) validation, and applying the corrective controller to individuals with SCI. Algorithm development consisted of AB individuals propelling a wheelchair with a wireless IMU across a variety of surfaces at a self-selected speed to record the accelerations. An algorithm was developed based on this preliminary data that could detect sudden stop events of the wheelchair while minimizing the false positives that could occur propelling over a variety of terrains. Once tuned, the controller was applied to three AB individuals recruited from a convenience sample of university students. AB subjects propelled a wheelchair over a pre-determined course near the institution to that contained the variety of surfaces and conditions expected in the community such as concrete sidewalks, paving bricks, curb cuts and transitions and locations that would cause sudden stops due to large potholes or curbs. After AB validation and troubleshooting, the controller was applied to three wheelchair users with SCI who received implanted stimulation systems with trunk and hip electrodes under other research protocols. All participants gave informed consent and all study procedures were approved by the local institutional review board. Each SCI subject acted as their own concurrent control with and without the sudden stop detection and fall prevention stimulation system active.

Algorithm development

To develop the control algorithm, acceleration data while propelling a wheelchair across a variety of surfaces were collected on two AB subjects. These surfaces included concrete sidewalks, paving bricks, curb cuts and transitions and locations that would cause sudden stops due to large potholes or curbs. A wireless sensor consisting of the accelerometer portion (± 8 g) of a LSM330DLC (ST Microelectronics; Geneva, Switzerland) IMU and a CC430F6137IRGC (Texas Instruments; Dallas, Texas) microcontroller with integrated 915 MHz wireless transceiver was placed on the rear cross bar of the wheelchair (Figure 1). The system sampled the tri-axial acceleration of the wheelchair at 50 Hz and transmitted it via the 915 MHz transceiver to an external control unit (ECU)[36] which implemented detection algorithms and issued commands to the implanted stimulation systems. A 3D printed housing mounted the sensor securely to the wheelchair and was lined with anti-vibration material to help minimize extraneous noise caused by normal propulsion over unchallenging surfaces.

To detect the large impulses seen in the collision trials while rejecting the noise from traveling over-ground over both smooth and rough terrain, a twenty-sample running root mean squared (RMS) value of the anterior-posterior (AP) acceleration was computed to reflect the energy of the signal, and act as a surrogate of the energy of the crash or the sudden stop. In order to help reject vibrations from traveling over uneven terrain, a baseline consisting of the moving average of the AP acceleration over the past two seconds was subtracted from the instantaneous readings. To ensure that the collision was sudden enough to warrant activation, the derivative of the RMS was compared against a threshold to indicate the instantaneous power of the signal (Figure 2). Thresholds were tuned using a custom Matlab (MATHWORKS, Natick, MA) program that adjusted the target thresholds for the RMS as well as its derivative. Data from the trials was run through a Matlab simulation of the controller to compare its outputs to true events. The thresholds were refined to achieve a greater than 95% accuracy of sample collisions while minimizing the number of false positives in the simulation. The optimal thresholds were then imported into the control algorithm implemented in the eternal control for real-time operation. The algorithm was then verified on three AB subjects propelling a manual wheelchair and three subjects with SCI. Figure 3 shows typical instantaneous

Figure 1 Wheelchair with wireless IMU affixed to rear axle (lid removed for picture).

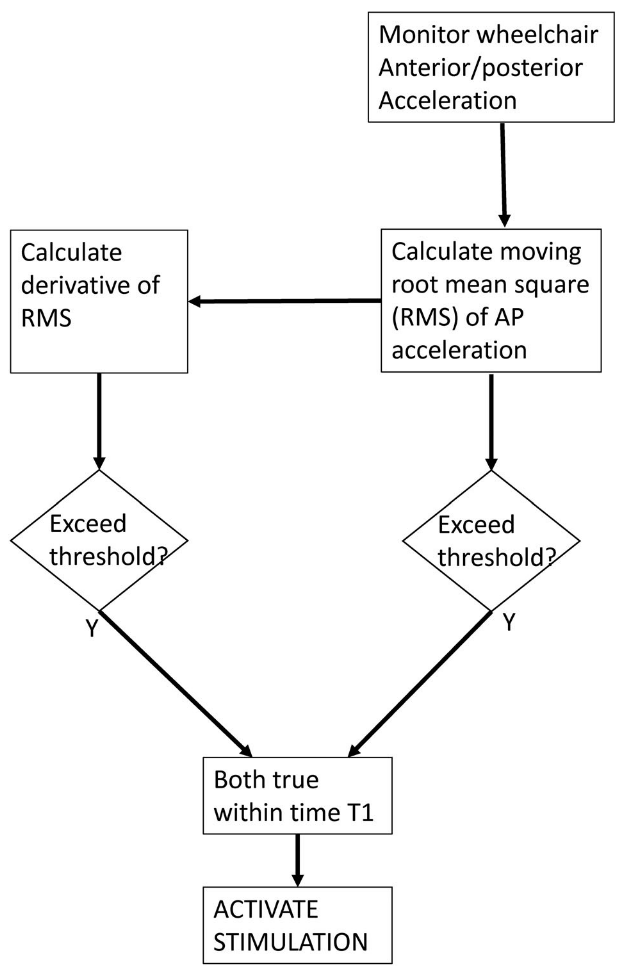

Figure 2 Block diagram of algorithm used to process acceleration signals and detect sudden stops.

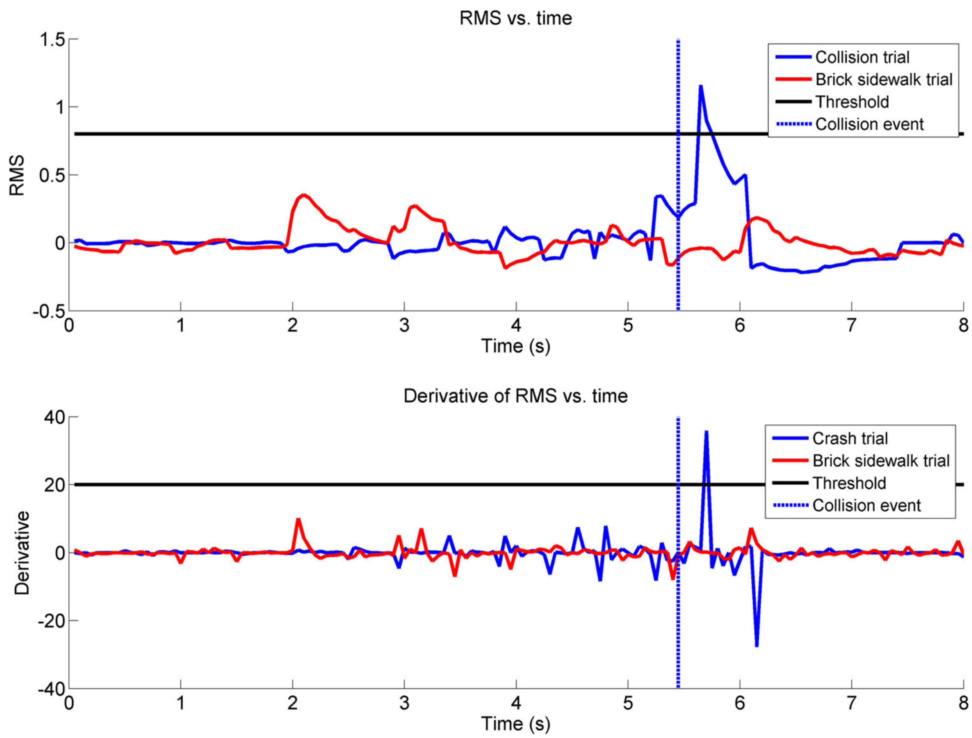

Figure 3 Plots of the RMS and derivative of the RMS during typical trials of two different conditions. The blue trace is a crash condition, which activated once the RMS and derivative signals crossed their respective thresholds. The red trace is an over-ground trial (no collision) traveling over a brick sidewalk. The activation thresholds are represented by the horizontal lines and the vertical dashed line indicates the time of the collision during that trial..

acceleration RMS and its derivative signals during propulsion over smooth surfaces and sudden stops, as computed from data sampled at 50Hz and down sampled to 20Hz to match the nominal stimulation.

Able-bodied testing

Three able-bodied volunteers (3 male, 22.3 ± 0.9 years old) propelled a manual wheelchair across various terrains. Each AB subject used a Top End T7A wheelchair and these subjects were not those involved in the preliminary data collection. The subjects were novice wheelchair users and were given approximately 5 min of practice propelling and to perform 5 pre-trial collisions to avoid learning effects. All conditions for each subject were performed in a single experimental session. Subjects were instructed to propel at a self-selected comfortable pace and a loose-fitting belt and standby guard was provided for safety. Subjects were instructed to propel at a self-selected comfortable pace and a loose-fitting belt and standby guard was provided for safety. Subjects propelled over a pre-determined route as described above in ***Study design***. All subjects rode over these surfaces for at least 10 min with the algorithm active. The external control unit was programed to beep and display a visual cue when collision events were detected, which were logged by an observer and compared to the actual presence or absence of a sudden stop. RMS and its derivative signals were collected during propulsion on each of the surfaces and stored in the memory of the ECU. The number of false positives (when the algorithm detected a collision when none happened) were recorded. The AB subjects also performed 20 sudden stop events along the route by colliding into objects, such as a curb, or lodging a castor in a pothole (Figure 4). Subjects were instructed to not brace for the collisions and study staff engaged the subject in conversation to provide distraction. The number of false negatives (when the algorithm failed to detect the actual collision) were collected and verified by the observer.

SCI testing

Concurrently with AB testing, three experienced wheelchair users with SCI propelled the same terrains using the same control algorithm. Each participant used his/her own manual wheelchair (Table 1) with no added suspension components and had previously received an implanted stimulation system for other standing/stepping,[37–39] seated stability,[31,33] or peripheral nerve cuff studies.[40–42] The systems consisted of an 8-channel implanted receiver-stimulator[43] or 16-channel

Figure 4 AB subject performing a sudden stop event (lodging castor into pothole).

implanted stimulator-telemeter[44] that was controlled via a transmitting coil connected to the ECU.

These systems utilized electrodes that delivered stimulating current to the nerves innervating the trunk and hip extensor muscles important for seated stability. Intramuscular electrodes[45] were inserted at L1-L2 level spinal nerves to recruit the lumbar paraspinal muscles for trunk extension. Two participants (S1, S2) had electrodes placed at T12-L1 to activate the quadratus lumborum for medial-lateral stability. Epimysial[46] or intramuscular electrodes were inserted or sutured at the motor points of the gluteal and hamstring muscles for hip extension. In some subjects, intramuscular electrodes were placed in the posterior portion of the adductor magnus to assist with thigh adduction as well as for hip extension. In one subject (S2), surface stimulation was applied for two muscles (right erector spinae and left quadratus lumborum) to augment the responses of the implanted electrodes. Subject demographics and the configuration of each system are summarized in Table 1.

The implanted devices delivered current controlled, charge-balanced pulses at 20 Hz and 20 mA, except for the paraspinal muscles in S1 and S2 in which the current amplitude was reduced to mitigate undesired co-activation of the abdominal muscles that can be observed with paraspinal stimulation.[47] Current amplitude was set at 100 mA for surface stimulation. Pulse width in all cases was independently varied (0–250µs) on a channel-by-channel basis and tuned for each subject. A nominal low level of stimulation was determined to ensure stability while propelling without restricting movement for each subject.[33] This was determined by tuning the pulse width of each muscle individually beginning at threshold (lowest value which produces a muscle contraction) and gradually increasing until it began to inhibit forward lean or produced an undesired movement (i.e. spillover to abdominals, hip rotation). A second, higher level of stimulation was then found that would to prevent forward trunk lean and return the user to an erect position after a collision or sudden stop.[48] This higher value was determined in a similar fashion by beginning at the low level stimulation value and gradually increasing each muscle until it reaches the hardware maximum (250µs) or begins to produce an undesired movement. This higher level of stimulation would then decrease back to the nominal values after 500 ms once the subject was assumed to have regained an upright sitting posture.

Subjects with SCI propelled their wheelchair across the same pre-determined route as the AB testing that consisted of a variety of terrains (concrete sidewalks, paving bricks, curb cuts and transitions) for between

Table 1 Subject demographics.

Subject	Sex	Level	AIS	Time post injury (y)	Time post implant (y)	Muscles stimulated	Wheelchair
S1	F	C7	C	20.86	8.39	BQL, BPA, BGM	Ti Lite TR 3
S2	M	C5	C	8.05	2.51	BES, BQL, BPA, BGM (RES and RQL were surface)	Ti Lite TR 3
S3	M	T4	B	11.11	7.25	BPA, BGM, BHS	Ti Lite ZRA
Mean				**13.34**	**6.05**		
StDev				*5.46*	*2.55*		

SCI subject demographics. Abbreviations: B = Bilateral, PA = Posterior portion of the adductor magnus, GM = Gluteus Maximus, HS = Hamstrings, ES = Erector Spinae, QL = Quadratus Lumborum.

twelve and fifteen minutes to determine the number of false triggers (Figure 5). Prior to each experiment, each subject propelled for approximately 3 min to ensure the algorithm did not have false positives with their wheelchair setup. The thresholds were minimally adjusted on the ECU to account for any setup differences. Each subject then performed 20 sudden stop events caused by hazards including large potholes in sidewalks and curbs with the application of corrective stimulation being applied randomized (ten of each condition with and without the control system active) in which the number of false negatives was determined. Prior to the experiment, each the 20 events were assigned to be a stimulation case or a non-stimulation case in a randomized fashion. Study staff activated the corrective stimulation on the ECU only during the assigned controller trials. Subjects were aware that stimulation would only be applied during certain trials, but were blinded to the experimental condition for each event. All conditions were performed in a single session. Users were instructed to propel at a self-selected pace and to not prepare or brace for the collisions. Efforts were made to distract users from anticipating the events by engaging in conversation and instructing them to look away from the path and at study staff. The Usability Rating Scale (URS),[49] a seven-point ordinal scale that ranges from "very difficult" (−3) to "very easy" (3) was administered after each collision to quantify subjective perceptions of safety with or without the control system active. For this study, the standard instrument was modified to ask the subject to rate each event "very unsafe" (−3) to "very safe" (3) as described in Triolo *et al.*[50] A loose-fitting seatbelt was worn around the waist for safety and subjects were spotted by a physical therapist providing stand-by assistance during the collision/sudden stop events.

Figure 5 Subject with SCI propelling over various terrains with algorithm active.

Data analysis

Algorithm accuracy for the AB and SCI testing was calculating by finding the percentage of controller detections as compared to the number of actual sudden stops. The number and rate of false positives (controller detections when there was no event) was calculated for both the AB and SCI participants.

The Wilcoxon signed-rank test was performed on the subjective rating data from each subject to check for statistical significance.

Results

Results from the able-bodied and SCI testing are shown in Tables 2–5. The accuracy of detecting the sudden stops (121 total for AB and SCI combined) from the wheelchair accelerations was 93.4% (Tables 2, 3). Detection success rates were consistently 95% for each SCI subject and varied between 90% and 95% for able-bodied participants. SCI subject S3 encountered one extra sudden stop during his over-ground testing when accidently colliding with a curb which is included in the Table 3.

All participants with SCI rated the event with corrective stimulation to be safer, with the results being significant for S1 (Z = −2.81, P = .005) and S3 (Z = −1.99, P = .046) (Table 3). All six participants propelled a manual wheelchair across various terrains for between

Table 2 Able-Bodied Detection Success Rate.

Subject	Collision Detection Success Rate
AB1	19/20 (95%)
AB2	18/20 (90%)
AB3	18/20 (90%)
Total	55/60 (91.7%)

Detection success rate for collision events with able-bodied subjects.

Table 3 SCI Users Detection Success Rate and URS Rating.

Subject	Collision Success Rate	URS (no controller)	URS (with controller)
S1	19/20 (95%)	−1 (barely unsafe)	2 (moderately safe)*
S2	19/20 (95%)	0 (neither)	1 (barely safe)
S3	20/21 (95%)	0 (neither)	1 (barely safe)*
Total	58/61 (95.1%)		

Detection success rate for collision events with median URS scores for SCI subjects. *indicates statistically significant change at P < .05.

Table 4 Able-Bodied False Triggers for Over-ground Propulsion.

Subject	Over-ground Conditions	Total Time	Number of False triggers	False Trigger Rate
AB1	Brick path, smooth concrete sidewalk, bumpy concrete sidewalk	19:40	2	1 every 9:50
AB2		21:36	3	1 every 7:15
AB3		22:15	2	1 every 11:08
Total		**63:31**	**7**	**1 every 9:04**

False Trigger rates for able bodied volunteers during free ranging over-ground propulsion over various terrains.

Table 5 SCI Users False Triggers for Over-ground Propulsion.

Subject	Over-ground Conditions	Total Time	Number of False Triggers	False Trigger Rate
S1	Brick path, smooth concrete sidewalk, bumpy concrete sidewalk	12:00	4	1 every 3:00
S2		14:00	2	1 every 7:00
S3		15:30	2	1 every 7:45
Total		**41:30**	**8**	**1 every 5:11**

False Trigger rates for SCI participants during free-ranging over-ground propulsion over various terrains.

twelve and twenty minutes. The mean number of false positives during these activities was 2.5 +/- 0.8 which corresponds to 1 every 6:56 min for the entirety of the 105 min tested over the six participants (Table 4).

Discussion

Individuals with paralysis who rely on wheelchairs for transportation encounter a variety of obstacles daily. Even as cities and sidewalks are updated and improved for access, they may still face obstacles that can challenge the stability of manual wheelchairs. This can include curbs, potholes, and gapped, cracked or uneven sidewalks that can cause their wheelchairs to stop suddenly. Such collisions and sudden stops can destabilize and dynamically alter the posture of the wheelchair user, which might ultimately result in potentially injurious falls.

This study demonstrates a method to detect the sudden stops which can cause excessive trunk lean in wheelchair users that may lead to seated instability and falls. By preventing the excessive trunk lean by stiffening the core via application of corrective stimulation to the hip and trunk extensors, the system can assist the user in maintaining a safe and stable posture when encountering either unexpected or anticipated sudden stops. Prior studies have shown how stimulation can significantly decrease the amount of trunk lean during a collision[35] and this study expands upon and extends those results to real-world community settings outside of the structured environment of the laboratory.

Detection accuracy of the algorithm presented was over 93% overall (91.7% for AB, 95.1% for SCI), which is comparable to motion-based detection and activity monitoring in wheelchair users.[51] When tuning the algorithm, it is vital to optimize the balance between false negatives and false positives and their effects on the user. A potential missed collision may cause the user to fall and decrease their confidence in the system so the detection system should be biased to minimize those false negative events. In other words, biasing the system toward false positives would be a safer option so as not to miss a potentially destabilizing event should it occur. However an excess of false positives, while not as dangerous, may be an annoyance to the user who may as a result elect to not use the system. The six users in this study propelled a wheelchair for a total of >100 min (63 for AB and 41 for SCI) and encountered 15 false positives (7 for AB and 8 for SCI), which sometimes surprised the users.

The detection accuracy of the system we developed and present could be further improved by incorporating additional sensors on the wheelchair or even mounted on the user to reduce false positive classifications. Expanding the sensor suite could include more information about the physical system into the decision-making algorithm, and potentially better discriminate between uneven, but not destabilizing, terrain and more challenging collisions or sudden stops. Furthermore, the system could be extended to detect rapid, tight-radius turns or other perturbations with other components of the linear acceleration vector or gyroscopic (angular velocity) signatures besides those confined to the sagittal plane in the anterior-posterior direction as in this study.

One limitation of this study may be the users' perceptions of safety and anticipatory compensatory actions prior to the collisions and destabilizing events. Although data were collected during sudden stops and challenging conditions in real-world settings, subjects still wore a loose-fitting safety belt and had stand-by assistance which would otherwise be unavailable in most home and community circumstances. The users could also see the upcoming obstacles, and although they were instructed to not alter their stroke patterns or prepare for the event in any way, they could have unintentionally braced themselves or made other postural corrections prior to encountering the sudden stops. Efforts were made to distract the subject with conversation and instruct them to continue looking upward during all trials. However, it was impossible to completely blind wheelchair users to the upcoming events. A future protocol could be developed where the users are unaware of upcoming challenges and the obstacles randomly placed in their propulsion path.

Although the sensor sampled and streamed the acceleration data at 50 Hz, the frequency of the stimulation was 20 Hz, which required the algorithm to down sample the output of the controller. An increase in sampling and processing rates may be of benefit since sudden stops could exhibit a sharper peak in acceleration that may not be captured adequately at slower frequencies.

Future work will include incorporating this system as an option in the take-home stimulation systems of implanted neuroprosthesis recipients. All of these participants have seated stability or wheelchair propulsion patterns in their take-home external control units that apply low-level constant stimulation while seated in a wheelchair. This detection scheme can be active in the background while their ECU is on and once an event is detected, the corrective stimulation would be applied to stiffen the hips and spine to prevent excessive forward or lateral lean in the event of a collision. This could then be expanded into a full terrain detection system that can distinguish between uneven surfaces, ramps, and sharp turns and apply the appropriate stimulation for whatever the condition may be.

The "neural seatbelt" system we describe can also be readily translated to surface stimulation to allow it to be more widely available to individuals without an implanted neuroprosthesis. Surface stimulation trunk control systems have been explored by various research groups.[29,52] This event detection and corrective action system can be extended to those surface stimulation-based systems since the detection algorithm is independent of the means for delivering neural activation. In addition, smart seat belts or other mechanical restraints could be integrated into similar destabilizing event detection systems for both manual and power wheelchairs. In those systems, a loose-fitting belt could be tightened or applied only when a destabilizing sudden stop happens. In this way, the user's static workspace would not be restricted when sitting still, and the constraints would only be applied when needed upon detection of an unsafe dynamic event.

Conclusion

A system to detect and act on potentially destabilizing events such as sudden stops during real-world conditions was developed, tested, and verified with manual wheelchair users with trunk and lower extremity paralysis. The system continuously monitored the root-mean-square and its derivative of the anterior-posterior acceleration of the wheelchair and applied corrective neural stimulation to activate the hip and trunk extensor muscles at a success rate of 95%. Users with SCI consistently rated the detection system as being safer than without event-triggered stimulation when undergoing sudden stops while propelling in a community setting. The number of false positive detections was minimized, although more investigation is needed to improve accuracy and ease implementation, as well as to generalize the results with a larger cohort of manual wheelchair users blinded to upcoming destabilizing events. This event detection and correction system can be integrated into the everyday take-home functions of implanted neuroprosthesis recipients to constantly monitor wheelchair status and apply corrective stimulation as needed. Such systems can also be made available to individuals without implanted devices via surface stimulation for trunk stability or mechanical restraints that apply lap or chest straps and belts only when necessary without compromising voluntary trunk motion during level propulsion over smooth surfaces or unduly affecting seated work volume during static sitting.

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